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(54) Optical coherence tomography assisted surgical apparatus

(57) An ophthalmologic surgical apparatus comprises a laser for producing laser radiation, laser delivery means for applying the laser radiation to an object and an optical coherence tomography ("OCT") apparatus. The ophthalmologic surgical apparatus has scanning means for scanning the object with optical output from the OCT apparatus, analysis means for analysing detection signals output from the OCT apparatus to determine portions of the object, which have been affected by the application of the laser radiation, and means, in response to output from the analysis means, for interacting with one or more of: (a) the laser, and (b) the delivery means to effect one or more of: (a) an exposure time, (b) a power, and (c) a spot size of the laser radiation.

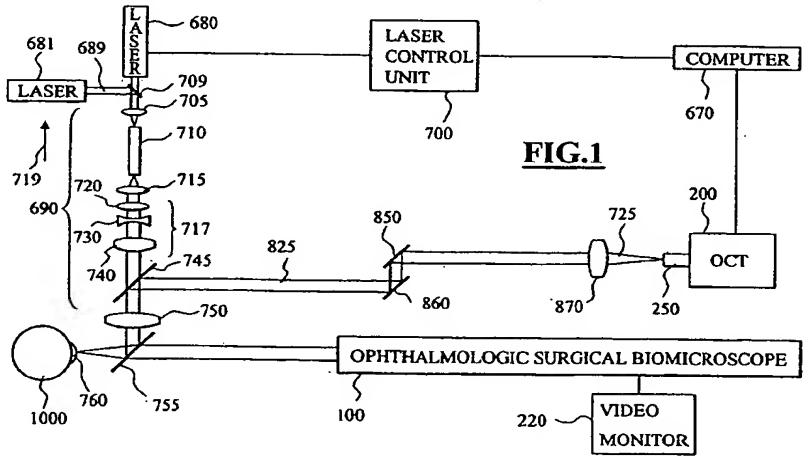

FIG.1

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Description

Technical Field of Invention

[0001] The present invention relates to a laser surgical apparatus which includes an optical coherence-tomography ("OCT") unit for controlling laser parameters such as exposure time, focus size, and power.

Background of the Invention

[0002] As is well known, cataract surgery is an ophthalmologic surgical procedure for removing an opaque intraocular lens from an eye. In accordance with this surgical procedure, after the intraocular lens is removed, an artificial intraocular lens needs to be implanted to recover the patient's vision. It ist desirable for an ophthalmologic surgical microscope that is used during the surgical procedure to have a capability of auto-focusing on the intraocular lens capsule during the surgical procedure, which capability is especially important after a majority of the opaque intraocular lens has been removed. After a majority of the opaque intraocular lens has been removed, small amounts of cataract residue may remain on the optically transparent intraocular lens capsule - because the intraocular lens capsule is transparent, such residue is difficult to see. As is known, it is important to completely remove such residue any residue left on the intraocular lens capsule will serve as a nucleus of a new cataract. Present apparatus for auto-focusing an ophthalmologic surgical microscope, such as a prior art apparatus disclosed in U.S. Patent No. 5,288,987 issued February 22, 1994, are based on detecting and measuring the intensity of light scattered from an object. However, such apparatus for auto-focusing are disadvantageous because it is difficult to focus on an optically transparent medium such as the posterior intraocular lens capsule since reflection therefrom is specular and weak.

[0003] In light of the above, there is a need in the art for an ophthalmologic surgical microscope which can auto-focus on the posterior intraocular lens capsule for use in cataract surgery.

[0004] As is well known, refractive surgery is a surgical procedure that has, as its primary objective, correction of an ametropia by making incisions in a cornea to change to refractive power of the cornea. Surgical manipulation of corneal shape requires an accurate and precise method of measuring anterior corneal curvature from apex to limbus. At present, measurement of curvature of the center of the cornea is commonly made using a keratometer and, for more precise measurements of corneal topography, it is common to utilize photokeratoscopy or videokeratoscopy.

[0005] Current corneal topography measurement apparatus are mostly Placido-disc-based videokeratoscopes. In such an apparatus, a series of concentric rings are configured on a cone-shaped housing so that an image reflected from the cornea is virtually flat in space. Then, the configuration of the rings is analyzed to determine the corneal topography. A prior art apparatus of this type has been described in an article entitled "New Equipment and Methods for Determining The Contour of the Human Cornea" by M. G. Townsley, Contacto, 11(4), 1967, pp. 72-81. Such videokeratoscopes have the following disadvantages: (a) due to the small radius of the cornea (~8mm), a limited number of rings can be resolved on the cornea (normally, the contour which can be measured is restricted to an area which ranges from 0.8 to 11 mm in diameter on the cornea); (b) no information can be obtained between the rings; and (c) due to use of rings, in-line measurement is very difficult when used in conjunction with an ophthalmologic surgical microscope. An article entitled "Accuracy and Precision of Keratometry, Photokeratoscopy, and Corneal Modeling on Calibrated Steel; Balls" by S. B. Hannush, S. L. Crawford, G. O. Waring III, M. C. Gemmill, M. J. Lynn, and A. Nizam in Arch. Ophthalmol., Vol 107, Aug. 1989, pp. 1235-1239 provides a comparison of these prior art methods and apparatus.

[0006] Another corneal topography measurement apparatus has been developed recently by PAR Microsystem Co. The apparatus utilizes raster photogrammetry to measure a corneal topography. In this apparatus, a grid pattern is projected onto the cornea. The grid pattern is then view and imaged from an offset angle. Finally, corneal elevation at each of the discrete points in the grid pattern are calculated using the image of the projected grid pattern, and information relating to its geometry. This apparatus is described in an article entitled "Intraoperative raster photogrammetry - the PAR Corneal Topography System" by M. W. Berlin, J. Cataract Refract Surg, Vol. 19, Supplement, 1993, pp. 188-192. Corneal topography measurements suffer in this apparatus because only a limited number of points in the image of the projected grid pattern can be resolved by the image optics.

[0007] As is further known, since a posterior corneal surface contributes about -14% of total corneal refractive power, in some cases, an anterior corneal topography, by itself, does not provide sufficient information for use in a refractive surgical procedure. For that reason, it becomes even more important to obtain, corneal topography measurements with a precision that cannot be provided by current corneal topography measurement apparatus.

[0008] In light of the above, there is a need in the art for an ophthalmologic surgical microscope which can perform in-line, corneal topography measurements for use in refractive surgical procedures.

[0009] Recently, a new ophthalmic measurement apparatus, an optical coherence tomography ("OCT") apparatus, has been disclosed which has advantages over the abovedescribed prior art ophthalmic measurement apparatus. An OCT apparatus uses a short coherence light source for range measurements based on the prin-

ciple of white light interferometry. OCT has been proposed recently for use in several ophthalmologic applications. For example, such proposals have been made in a preprint of an article which has been submitted for publication entitled "Micron-Resolution Imaging of the Anterior Eye in Vivo with Optical Coherence Tomography" by J.A. Izatt, M. R. Hee, E. A. Swanson, C. P. LIN, D. Huang, J. S. Schumann, C A. Puliafity, and J. G. Fujimoto, 1994, pp. 1-24. The preprint discloses an OCT apparatus which utilizes optical fiber technology and a superlumineascent laser diode source, which OCT apparatus is interfaced with a slitlamp biomicroscope for imaging intraocular structures with a spatial resolution of 10-20 μm.

[0010] The preprint discloses the use of the OCT apparatus to procide direct, micron-resolution measurement of (a) ocular profile dimensions, optical scattering, and structure in the cornea; (b) the anterior angle region; (c) the iris; and (d) the crystalline lens. The preprint further discloses the use of the OCT apparatus to measure: (a) anterior chamber depth, defined as a distance, along the visual axis, from the posterior corneal surface to the lens anterior capsule (b) radius of curvature of the posterior and anterior surface of the cornea; (c) corneal refractive power; and (d) corneal dimensions such as thickness. The preprint still further discloses that the OCT apparatus, using an inexpensive diode laser source and a fiber optic implementation, is compatible with existing ophthalmic instrumentation. Finally, the preprint makes the following suggestions for potential clinical applications of OCT: (a) providing cross-sectional images of the entire anterior chamber for use in elucidating pathologies of the cornea, anterior angle region, and iris and for use in identifying and monitoring intraocular masses or tumors; (b) measuring anterior chamber depth, corneal curvature, and corneal refractive power; and (c) providing high resolution image showing corneal thickness variations and the distribution of scattering in corneal stroma for quantitative analysis of corneal pathologies.

[0011] As is well known, lasers are used in eye surgery for various applications, of which, perhaps the most important are photocoagulation of the retina and photoablation of the cornea. In such applications, laser radiation interacts with ocular tissue and causes structural and topological changes of the tissue. Such applications typically entail monitoring such tissue changes visually on a video monitor by means of a CCD microchip interface or through a binocular eye piece with an ophthalmologic surgery biomicroscope. However, the CCD image of the prior art is limited for two basic reasons. The first reason the CCD image of the prior art is limited is that the CCD image only provides an image of tissue surface. For laser treatment of macular holes, for example, although there is a need to limit tissue coagulation to a well defined area to avoid unnecessary damage of visual functions, there is also a need to limit tissue coagulation in depth to avoid bleeding of the highly perfused coroidal layer. Another example of the need to limit tissue changes in depth is the need to avoid damage of the endothelium layer of the cornea during laser ablation for photorefractive surgery. The second reason the CCD image of the prior art is limited is that is does not provide a quantitative method for controlling tissue change based on laser power, exposure, and spot size.

[0012] In light of the above, there is a need for an apparatus for use in laser treatment for controlling the extent of tissue change during the laser treatment and for controlling the tissue change based on laser power, exposure, and spot size.

Summary of the Invention

[0013] An embodiment of the present invention comprises a laser delivery system which is combined with and OCT apparatus wherein the extent of tissue change during laser treatment is monitored and controlled by analyzing output from the OCT apparatus to control the laser delivery system.

Brief Description of the Figures

[0014]

FIG. 1 shows, in pictorial form, an embodiment of the present invention for monitoring and controlling the extent of tissue change during laser treatment comprising an optical coherence tomography ("OCT") apparatus:

FIG. 2 shows, in pictorial form, a fiber optic embodiment of the OCT apparatus shown in FIG. 1.

[0015] Components which are the same in the various figures have been designated by the same numerals for ease of understanding.

Detailed Description

[0016] FIG. 1 shows, in pictorial form, an embodiment of the present invention which comprises optical coherence tomography ("OCT") apparatus 200, computer 670, treatment laser 680, aiming laser 681, laser delivery optics 690, laser control unit 700, ophthalmologic surgical biomicroscope 100, and video monitor 220 as video imaging unit.

[0017] The ophthalmologic surgical microscope 100 is comprised of objective lens which has a long working distance (~200 mm) for focusing on patient's eye 1000 during a surgical procedure. The ophthalmologic surgical microscope 100 further comprises optical magnification changer which is set to a condition suitable for performing a particular surgical procedure (typically there are a number of groups of lenses arranged on a drum for providing varying magnifications such as, for example, 5X, 12X, 20X, and so forth). Radiation imping-

ing upon the optical magnification changer is collimated.

[0018] Ophthalmologic surgical microscope 100 further comprises: (a) relay lenses which take collimated radiation output from an optical magnification changer and form an intermediate image of an object, for example, eye 1000; and (b) internal focusing lenses which are used to focus on the intermediate image of the object formed by relay lenses and provide a collimated beam (internal focusing lenses move up and down along viewing path to provide an opportunity for internal focus adjustment).

[0019] After passing through internal focusing lenses, radiation is collimated and a beamsplitter couples a portion of the collimated radiation into an optical path for obtaining a video image. The video image is obtained by use of a video lens, a CCD camera, and video monitor 220. As those of ordinary skill in the art can readily appreciate, although the use of a single CCD camera is shown, it is within the spirit of the present invention that embodiments may be fabricated utilizing two beamsplitters, i. e, a first beamsplitter and a similarly placed second beamsplitter, to provide stereoscopic viewing through two CCD cameras.

[0020] Lastly, tube lenses focus collimated radiation passed through beamsplitters at an object plane of eye pieces. These eye pieces then provide collimated output which is focused by a viewer's eyes. The above-described viewing path is binocular, therefore stereoscopic viewing can be obtained.

[0021] Output from treatment laser 680 passes through dichroic beamsplitter 709 and is focused by lens system 705 into lightguide 710 and output from aiming laser 681 is reflected by dichroic beamsplitter 709 and is focused by lens system 705 into lightguide 710. Embodiments of treatment laser 680 for use in, for example, photocoagulation and photoablation, are well known in the art and embodiments of aiming laser 681 are also well known in the art. For example, one well known laser used to embody aiming laser 681 is a He-Ne-Laser.

[0022] The output from lightguide 710 is converted into a substantially parallel laser beam by collimating lens 715, which substantially parallel laser beam is applied as input to parfocal system 717. As shown in FIG. 1, parfocal system 717 is comprised of converging lens 720, diverging lens 730, and converging lens 740. Lens 720 and 730 are movable along the axis of laser delivery optics 690. Apparatus for moving lenses 720 and 730 along the axis of laser delivery optics 690 are well known in the art and are not shown for clarity and for ease of understanding the present invention. The direction of the axis of laser delivery optics 690 is shown by arrow 719 and will be referred to below as the z-axis. As is well known in the art, the output from parfocal system 717 is a substantially parallel beam having a variable beam diameter, the size of the beam diameter being determined by the positions of lenses 720 and 730 along the z-axis, i.e., the axis of laser delivery optics 690.

[0023] As shown in FIG. 1, the output from parfocal system 717 passes through beamcombiner 745 and is applied as input to micromanipulator lens 750. The output from micromanipulator lens 750 is directed by beamdirector 755 to laser focus 760 on eye 1000. FIG. 1 shows the laser beam being focused onto the cornea of eye 1000. However, it should be clear to those of ordinary skill in the art that the laser beam can also be focused onto the retina with the same configuration by, for example, using a contact lens which is pressed against the cornea of the patient's eye. The purpose of the contact lens is to neutralize the refractive power of the cornea. In this case, the working distance between the eye and the apparatus is adjusted so that the laser beam is focused onto the retina.

[0024] Beamcombiner 745 combines laser radiation output from parfocal system 717 with radiation from OCT beam 725 which is output from OCT apparatus 200. In a preferred embodiment, beamcombiner 745 is a beamsplitter which transmits laser radiation at wavelengths of the output from treatment laser 680 and aiming laser 681 and reflects radiation at wavelengths of OCT beam 725. In the preferred embodiment, beamdirector 755 is a mirror or a reflecting prism which is located between the two observation paths of ophthalmologic surgical biomicroscope 100. Micromanipulator lens 750 is movable in a plane which is perpendicular to the z-axis, which plane will be referred to below as the x-y plane. Apparatus for moving micromanipulator lens in the x-y plane are well known in the art and are not shown for clarity and for ease of understanding the present invention.

[0025] As shown in FIG. 1, OCT beam 725 is output from fiber 250 of OCT apparatus 200 and is collimated by lens 870. The collimated output from lens 870 impinges upon scanning mirrors 850 and 860 which are orthogonally mounted, galvanometer driven scanning mirrors. Scanning mirrors 850 and 860 are mounted on a pair of motors (not shown) which are operated under the control of computer 670 in a manner which is well known to those of ordinary skill in the art to provide transverse scanning of OCT beam 725. OCT beam 825 which emerges from scanning mirrors 850 and 860 is directed towards beamcombiner 745. In accordance with the present invention, scanning mirrors 850 and 860 are located substantially at the back focal plane of micromanipulator lens 750.

[0026] Scanning mirrors 850 and 860 are driven with a sawtooth profiled voltage function in a manner which is well known to those of ordinary skill in the art. When the phase and frequency of the respective driver voltages are equal, the resulting scan pattern is a Linear scan.

[0027] FIG. 2 shows, in pictorial from, a fiber optic embodiment of OCT apparatus 200. As shown in FIG. 2, OCT apparatus 200 comprises CW radiation source 220, for example, a superluminescent laser diode having an output centered substantially at 850 nm. Output from source 220 is coupled into optical fiber 230 and is separated into two beams by 50/50 coupler 240. The

output from 50/50 coupler 240 is coupled into optical fibers 250 and 270, respectively. The output from fiber 270 is imaged by lens 280 onto reference mirror 290 and output from fiber 250 is directed to transverse scanning mechanism 260. The output from transverse scanning mechanism 260 is directed to impinge upon an object in a manner to be described in detail below. Then, radiation reflected from the object is coupled back into fiber 250 and superimposed by 50/50 coupler 240 with radiation reflected from reference mirror 290 and coupled back into fiber 270. Superimposed radiation output from 50/50 coupler 240 is coupled into fiber 265. As is known, there is interference between radiation reflected from the object and radiation reflected from reference mirror 290 if the optical path difference is smaller than the coherence length of radiation source 220. Reference mirror 290 is moved with a substantially constant velocity by means which are well known to those of ordinary skill in the art (not shown) and, as a result, the interference is detected as a periodic variation of a detector signal obtained by photodetector 275, the periodic variation having a frequency equal to a Doppler shift frequency which is introduced by moving reference mirror 290 with the constant velocity. The output from photodetector 275 is demodulated by demodulator 285, the demodulated output from demodulator 285 is converted to a digital signal by analog-to-digital converter 295 (A/D 295), and the output from A/D 295 is applied as input to computer 670 for analysis. The interference signal vanishes as soon as the optical path difference between radiation reflected from the object and radiation reflected from reference mirror 290 becomes larger than the coherence length of source 220.

[0028] In a preferred embodiement, the OCT beam has a wavelength centered about 850 nm.

[0029] Referring to FIG. 1, in a preferred embodiment, to OCT beam has a wavelength entered about 850 nm, the linear scan (produced by scanning mirrors 850 and 860) combines with an OCT scan in a longitudinal direction into eye 1000 (produced by movement of reference mirror 290 shown in FIG. 2) to provide an OCT scan in a plane. The amplitudes of each sawtooth profile can be individually adjusted, in a manner which is well known to those of ordinary skill in the art, to change the orientation of the linear scan and, hence, the orientation of the plane of the OCT scan. The orientation of the linear scan and, hence, the orientation of the plane, are determined by the ratio of the amplitudes of each sawtooth profile. Thus, in accordance with the present invention, the plane of the OCT scan can be rotated (the z-axis is the axis of rotation) by varying the ratio of the amplitudes.

[0030] In the apparatus shown in FIG. 1, the principal rays of OCT beam 825 are in the same plane as the laser beam emerging from laser delivery system 690. Thus, OCT beam 825 is focused in the same focal plane as the beams from treatment laser 680 and aiming laser 681. In additon, OCT beam 825 moves when micromanipulator lens 750 is moved in the x-y direction. As a result, OCT beam 825 always scans transversely across laser focus 760 in the object plane and scans transversely across laser focus 760 symmetrically, i.e., the length of the portions of the transverse scan on either side of laser focus 760 are of equal length.

[0031] Radiation from OCT beam 725 is backscattered from eye 1000 and is coupled back to OCT apparatus 200. As described above with respect to FIG. 2, an OCT signal is generated by OCT apparatus 200. This signal is sent to computer 670 for analysis. Further, radiation from aiming laser 681 is reflected from eye 1000 through beamdirector 755 and into ophthalmologic surgical biomicroscope 100. Finally, video monitor 220 provides an image which is used to position laser focus 760 by changing the position of micromanipulator lens 750. The ophthalmologic surgical microscope 100 provides an image of eye 1000 for viewing in conjunction with the position of laser focus 760 on video monitor 220.

[0032] The following describes the analysis performed by computer 670 to control the operation of treatment laser 680. The amplitude of back scattered radiation from OCT beam 825 is different for laser-treated tissue and for untreated surrounding tissue. In fact, experimental studies show that the amplitude of such back scattered radiation is higher for laser-treated than it is for untreated surrounding tissue and this fact is used to identify laser-treated tissue.

[0033] The reflectivity of laser-treated tissue and untreated tissue is determined empirically as a function of, for example, laser power and exposure time. These data are utilized to develop reference threshold levels for use in the analysis. For example, an amplitude having a value above the reference threshold level is deemed to have been received from laser-treated tissue and an amplitude having a value below the reference threshold level is deemed to have been received from untreated tissue. As those of ordinary skill in the art understand, the effects of noise and small movements of the eye must be taken into account when determining the reference threshold levels. These reference threshold levels are stored in computer 670. As one can readily appreciate, the reference threshold levels may be a function of the type of tissue involved and/or the laser power and/or the exposure time. However, for purposes of understanding the present invention, we will assume that there is a single reference threshold level. With the following explanation, it should be clear to those of ordinary skill in the art how embodiments may be fabricated to take into account more complex variations of the reference threshold level.

[0034] In accordance with the present invention, computer 670 activates OCT apparatus 200 and scanning mirrors 850 and 860 to provide an OCT scan of a portion of eye 1000. As was described above, scanning mirrors 850 and 860 produce a linear, transverse OCT scan and, at predetermined points in the linear, transverse OCT scan, OCT radiation reflected from all scatterers

in the path of OCT radiation 825, along a longitudinal direction into eye 1000, is compared with radiation from a reference path whose optical length is varied periodically and which optical length is accurately known (see the description of OCT given above in conjunction with FIG. 2). As was described above, an OCT output signal is generated only when the optical length of the path of OCT radiation reflected from a feature of eye 1000 is equal to the optical length of the reference path, to within the OCT radiation temporal coherence length. In accordance with the present invention, amplitude information is obtained for reflected OCT radiation, as a function of depth along a longitudinal direction into eye 1000, at each of the predetermined points in the linear, transverse scan. Thus, after the OCT scan, computer 670 has collected data which comprises amplitude information over a plane (referred to below as a transverse plane), the transverse plane extending symmetrically about laser focus 760 and along the longitudinal direction into eye 1000. The data are analyzed by computer 670 to determine the area of the laser-treated tissue within the transverse plane. The area is compared with information regarding a desired clinical effect in a predetermined regimen of treatment, which information has been input to computer 670, for example, by a physician. Computer 670 then determines whether further exposure is warranted to achieve the desired clinical effect or whether the desired effect has been achieved. Further, computer 670 interacts with treatment laser 680 by means of laser control unit 700 to turn off treatment laser 680 or to change the exposure time. Still further, computer 670 can indicate the need to change the spot size and can provide for positioning of lenses 720 and 730 to achieve a desired spot size on a display which is a part of computer 670 so that an operator can position lenses 720 and 730 appropriately. Alternatively, lenses 720 and 730 can be mounted on motors (not shown), for example, stepper motors, and computer 670 can cause lenses 720 and 730 to be moved to provide the desired spot size by direct interaction with the motors in a manner which is well known to those of ordinary skill in the art. Another embodiment of the analysis is carried out by setting the reference threshold level to be a predetermined fraction of the maximum amplitude within the OCT image. The predetermined fraction is empirically determined. This embodiment of the analysis is less sensitive to variation of the backscattered intensity due to variations of the transmission of the various sections of the eye from patient to patient.

[0035] Laser-treated tissue is identified and the laser treatment process is controlled in accordance with the present invention as follows. Data from a transverse plane are obtained during pulses of treatment laser 680 and the data are stored in computer 670. A typical transverse plane includes an area of about 2 mm by 1 m and a typical set of data for the typical transverse plane includes data from a grid of about 30 by 100 values. The length of a typical laser pulse used in treatment is about 100 ms laser pulse. This means that data from each transverse plane (comprised of 3000 values) has to be processed in 10 ms. In the preferred embodiment, in analyzing the amplitude data for a transverse plane, if the amplitude at a location in the grid is higher than the reference threshold level, it is assumed that grid location corresponds to laser-treated tissue. As the data for each location in the grid is analyzed, an entry is made in a matrix corresponding to the location in the grid- of laser-treated tissue and a counter is increased by one step for each location at which laser-treated tissue is detected. The counter counts the number of grid locations in the transverse plane where the effects of laser treatments are measurable. The process is completed after a full OCT data set for the transverse plane comprising 30 by 100 data values has been acquired. The final counter value gives a measure of the laser treated area of the transverse plane. In a further embodiment of the present invention, the data for the grid is imaged on the display in such a manner that grid locations which have been determined to show the effects of laser treatment are identified. For example, the values which exceed the reference threshold may be displayed in a different color from that of surrounding tissue. This provides a visual picture of the effects of the laser treatment over the transverse plane for the operator.

[0036] In the embodiment described above, it is advantageous to analyze the data from a transverse plane to save processing time. This advantages occurs when using a symmetrical laser beam so that the effect of the laser treatment is symmetrical about a line along the longitudinal direction into eye 1000. However, the present invention includes embodiments which detect laser-treated tissue over a volume. In accordance with the present invention, data for a volume of eye 1000 is obtained by rotating the direction of the linear, transverse OCT scan by a predetermined amount in the manner described above, i.e, by varying the ratio of the amplitudes of the sawtooth voltages applied to drive scanning mirrors 850 and 860. Then, data are collected for the plane at the rotated position, den plane is rotated again, and so forth, until data is collected for the volume. The data are analyzed in the manner described above to provide a matrix which identifies laser-treated tissue. In addition, counters for all of the transverse planes are summed to provide a measure of the laser-treated volume. In the further embodiment, the volume is imaged on the display in such a manner that locations which have been determined to show the effects of laser treatment are identified. For example, the values which exceed the reference threshold may be displayed in a different color from the that of surrounding tissue. This provides a visual picture of the effects of the laser treatment over the volume.

[0037] In accordance with the present invention, the measured area of a transverse plane of laser-treated tissue is used to control one or more of the following parameters: (a) for treatment laser 680: exposure time and

power and (b) for lenses 720 and 730; the spot size of treatment laser 680. As described above, the measured area of a transverse plane of laser-treated tissue is measured several times during a laser exposure and the measured areas are compared with empirically determined reference values for each of those measurements. In a preferred embodiment, the reference values are selected so that the area of the laser-treated tissue in a transverse plane is small enough to avoid damage to neighboring tissue and, in accordance with the present invention, the laser is automatically turned off before this limit value is reached. As one can readily appreciate, the exposure time and power for treatment laser 680 and the spot size of the beam may be found to depend on the difference between a measured amplitude and a reference threshold level. In such cases, empirical studies may be used to control the values of these parameters depending, not only on whether an amplitude exceeds or falls below a reference threshold level, but on the value of the difference between the amplitude and the reference threshold value. It should be clear as to fabricate embodiments of the present invention which take such effects into account in light of the above.

[0038] In further embodiments, a measured volume of laser-treated tissue is used to control one or more of the following parameters: (a) for treatment laser 670; exposure time and power and (b) for lenses 720 and 730; the spot size of treatment laser 670. In accordance with the present invention, the measured volume of laser-treated tissue is measured one or more times during and the measured volumes are compared with empirically determined reference values for each of those measurements. It should be understood that the detection of areas and/or volumes of laser-treated tissue are not limited to detection during the application of a laser pulse and that such detection may take place after the laser pulse has been applied.

[0039] Those skilled in the art will recognize that the foregoing description has been presented for the sake of illustration and description only. As such, it is not intended to be exhaustive or to limit the invention to the precise form disclosed.

Claims

1. Ophthalmologic surgical apparatus which comprises:

   a laser for producing laser radiation;
   laser delivery means for applying the laser radiation to an object;
   an optical coherence tomography (OCT) apparatus (200);
   scanning means (850, 860) for scanning the object with optical output from the OCT apparatus (200);
   analysis means (670) for analyzing detection signals output from the OCT apparatus to determine portions of the object, which have been effected by the application of the laser radiation; and
   means (700), in response to output from the analysis means (670), for interacting with one or more of: (a) the laser (680), and (b) the delivery means (720, 730) to effect one or more of: (a) an exposure time, (b) a power, and (c) a sport size of the laser radiation.

2. The ophthalmologic surgical apparatus of claim 1 wherein the analysis means comprises means for comparing the detection signals output from the OCT apparatus with reference threshold levels.

3. The ophthalmologic surgical apparatus of claim 1 or 2 wherein the analysis means further comprises means for determining the portions as one or more of: (a) areas of transverse planes of the object and (b) volumes of the object.

4. The ophthalmologic surgical apparatus as in one of the preceeding claims wherein the scanning means comprises means for scanning the optical output from the OCT apparatus across a focus of the laser radiation.

5. The ophthalmologic surgical apparatus as in one of the preceeding claims wherein the means for scanning across the focus comprises means for scanning symmetrically across the laser focus.

6. The ophthalmologic surgical apparatus as in one of the preceeding claims wherein the analysis means comprises means for counting the number of instances in which the detection signals exceed threshold levels for one or more areas of transverse planes of the object and for comparing the number for the one or more areas with stored treatment data.

7. The ophthalmologic surgical apparatus as in one of the preceeding claims wherein the portions of the object which have been effected are determined to be portions wherein the detection signals exceed the reference threshold levels.

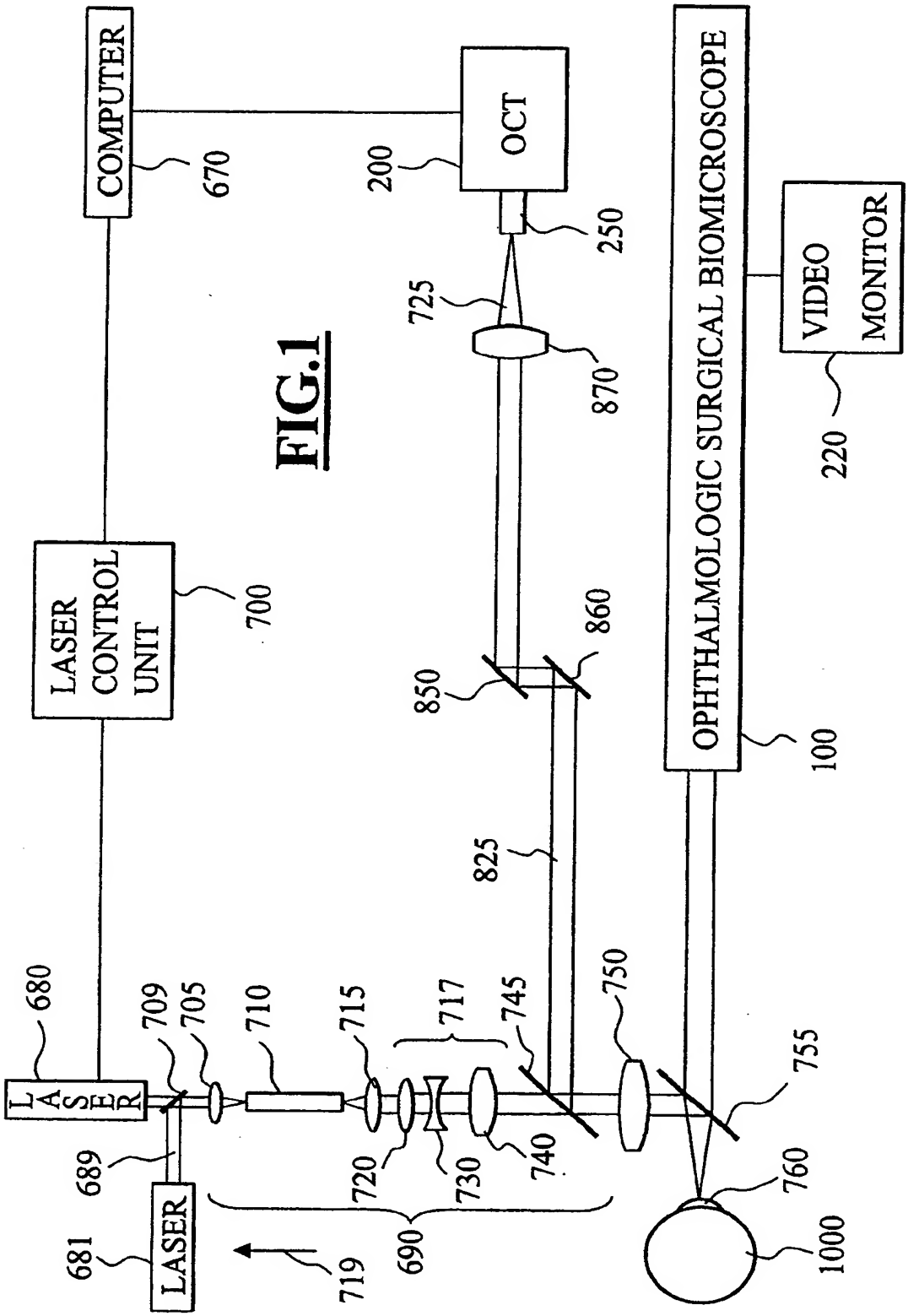
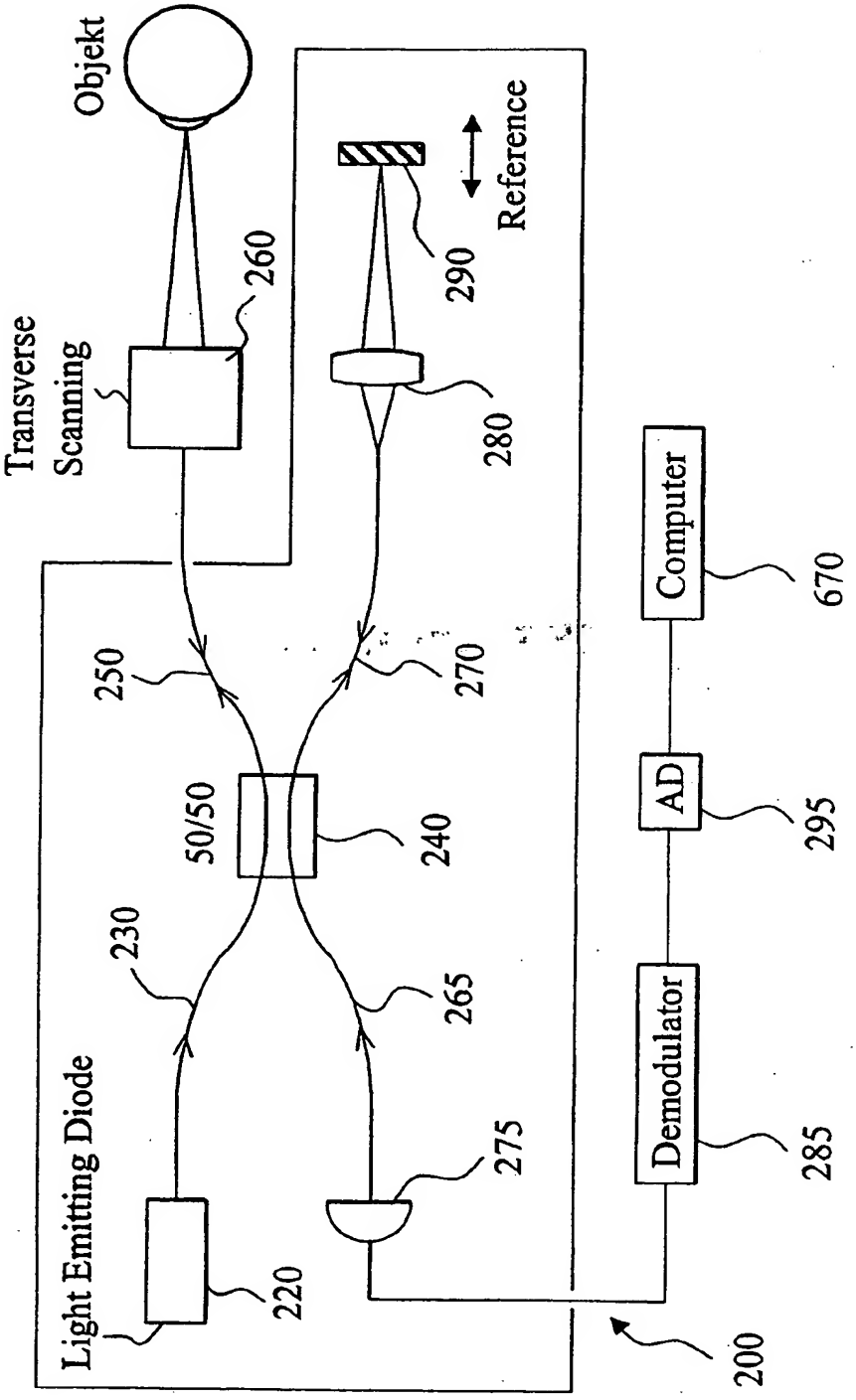


FIG.1

FIG.2



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(54) Optical coherence tomography assisted surgical apparatus

(57) An ophthalmologic surgical apparatus comprises a laser for producing laser radiation, laser delivery means for applying the laser radiation to an object and an optical coherence tomography ("OCT") apparatus. The ophthalmologic surgical apparatus has scanning means for scanning the object with optical output from the OCT apparatus, analysis means for analysing detection signals output from the OCT apparatus to determine portions of the object, which have been affected by the application of the laser radiation, and means, in response to output from the analysis means, for interacting with one or more of: (a) the laser, and (b) the delivery means to effect one or more of: (a) an exposure time, (b) a power, and (c) a spot size of the laser radiation.

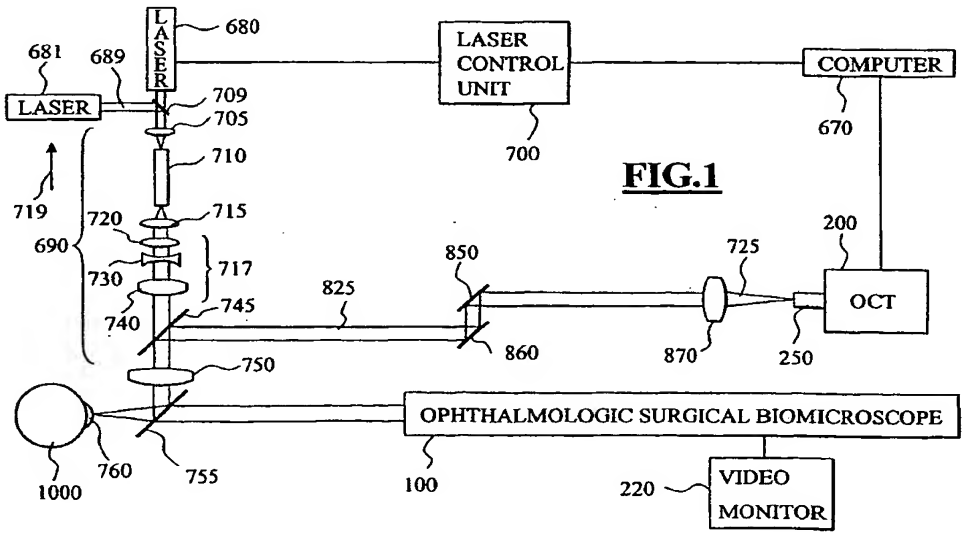

FIG.1

EP 1 231 496 A3

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EUROPEAN SEARCH REPORT

Application Number
EP 02 00 5702

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The present search report has been drawn up for all claims

Place of search	Date of completion of the search	Examiner
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EPO FORM 1503 03.82 (P04C01)

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